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DESIGN OF A COMPLIANT MECHANISM BASED PROSTHETIC FOOT

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ABSTRACT

The expensive prosthetic foot products available today are unaffordable to the amputees below the low income bar. In this paper, our methodology to develop a low cost Compliant Mechanism based Prosthetic Foot is highlighted. The base design of our foot is obtained by the execution of the Topology Optimisation code followed by the development of a 3D CAD model of the required dimensions. The typical pressure line of an average human foot (60 Kg) is considered and incorporated into the CAD model to endorse flexibility, efficient energy storage and return (ESAR) when compared to the conventional behavior of rigid prosthetic foot products. Finite Element Analysis (FEA) is further carried out discreetly, accounting three major phases the human foot undergoes during a gait cycle.

KEYWORDS: Compliant Mechanism, Topology Optimisation, Prosthetic Leg & FEA

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Original Article

INTRODUCTION

The prosthetic market today is providing solutions in an ineffective manner in order to help the amputees maneuver the routine challenges. This is primarily due to the fact that there are innumerable limitations involved in the interface adapted for controlling the prosthesis and a dearth of tactile feedbacks, limiting the capabilities [1]. With robust design inputs in the future, we can achieve greater satisfaction levels from the amputee's side. Most people with lower-limb amputation (LLA) rely on the prosthetic foot to support them over the course of their lifetime. Fogelberget al focused on documenting the accounts of peoples' experiences with the prosthetic devices [2]. This study, hence increased the understanding of the impact the current devices can deliver. Design engineers can obtain a clear insight on future materials and fabrication methods required to develop prosthetic products in the future. The current amputees report high confidence levels in forward walking on even surface, suggesting that current prosthetic foot products are competent in achieving their primary objective. However, a number of situations such as walking on sidewalks filled with rocks, climbing ladders and stools were identified and inferred that they continue to pose a difficulty for the patients. Social discomfort and body-image anxiety has been found among people with amputation [3]. These apprehensions in the minds of patients coupled with increased activity restrictions, which further deepen the depression levels. LLA patients generally require a longer period of time to cope with the amputation, garnering greater social support and developing an optimistic personality temperament. The majority of studies on adjustment to amputation is cross-sectional in design and have used non-comparable procedures. Additionally, they have neglected to study many important areas of prosthetics, including immediate reaction to amputation, adjustment during and shortly after the rehabilitation period and most importantly to develop a changed sense of self and identity. In order to discuss these concerns, a cohesive approach is required from researchers and engineers, thus leading to a holistic development of prosthetic foot that is affordable to a wide

range of patients across the income bar. Patients in the low income bar can hence easily afford prosthetic products within their grasp, contrary to heavy investment on an expensive prosthetic foot powered by actuators and sensors.

It is obligatory to ensure that a developed prosthetic foot suits the requirement of a general sample of the population, thus promoting ease of adjustment [4]. The past research in prosthetics always focused on restoring the mechanical energy properties in the ankle joint region of the developed foot. Takahashi et al has devised a way to quantify the total mechanical energy profiles of a natural human foot and hence conclude that a prosthetic device developed in the future must exploit the Energy Storage and Return (ESAR) principle, hence imitating the natural energy profiles of the natural ankle foot system during walking on uniform surfaces. It is hence of paramount importance that we consider a large proportion of factors in order to prevent damaging the lower limbs. Factors leading to limb loss consist of: reduced walking speeds [5], altered residual leg muscle patterns [6], asymmetry in ground force reaction patterns [7], higher metabolic costs [8], irregular joint loading at the knee region [9], chronic back and leg pain [10]. The study of basic biomechanics of the foot plays a major role in the ascertaining the orientations of the foot in different gait cycles [11]. A three-dimensional finite element analysis (FEA) model, including cartilage and ligaments were modeled and simulated for a typical human gait cycle. A gait cycle consists of three phases: heel strike, midstance and toe lift off conditions. The role of individual muscles during different phases of gait cycle were also inferred. Tibialis anterior, Extensor halluces longus and tibialis posterior, flexor tendons contribute to flexibility of the human foot. Another major gap in the existing design is the non-consideration of Achilles tendon. Its major function is to accord the calf muscle (planter is) to heel bone (calcaneus). Plantar flexion of the foot at the point of the ankle is mainly achieved through the support of this tendon.

Contemporary Prosthetic Foot Models

Segmented foot with two series elastic actuators (one for the ankle joint and other for toe joint) was designed by Wang et al [12]. Aluminum alloy was considered as primary material for fabrication. Each segmented foot comprised of a DC motor, a ball screw transmission and a spring structure. The design of the powered joints was based on the functionality of human toe and ankle joints. The gait cycle of the foot resembled natural human walking due to the rotation around toe joint, toe tip. Passivity based dynamic bipedal model had been proposed to analyze the energy efficiency and the stability of bipedal walking. The effects of foot structure on motion characteristics such as energy efficiency and walking stability is investigated through simulation experiments. The energy consumption during walking was further tabulated. Fey et al generated the final design employing topology optimization technique [13]. The optimization algorithm solved for keel section geometry by minimizing volume and attempting to match the compliance of the keel section of the nominal carbon fiber foot in a toe-only loading condition. Then, the previously-optimized geometry of the keel section was fixed, and the shape of the heel section was optimized to minimize volume and match the carbon fiber compliance in a foot-flat loading condition. This step was followed by the creation of CAD models and analysis through the FEM package. FEM simulations were performed to investigate the material usage and loading responses at different instances. The prototype was fabricated using Selective Laser Sintering (SLS) technique.

METHODOLOGY

The goal of this study was to develop a framework implementing topology optimization techniques to generate a CAD model and employ FEA Analysis to predict the behavior of the foot in realistic conditions, well-suited for an array of applications.

Topology optimization

Topology optimization has been a widely used technique which effectively promises to optimize the layout of a product component [14]. In medical field applications, compliant mechanisms optimized by this technique have found a range of applications in designing amputee prosthetic sockets [15], stimulating bone growth [16]. Topology optimization techniques to yield revolutionary prosthetic leg designs is an emerging field. Topology optimization is a mathematical approach that optimizes material, layout within a given design space, for a given set of loads and boundary conditions such that the resulting layout meets a prescribed set of performance targets. Using topology optimization, engineers can find the best concept design that meets the design requirements. The required design domain is represented in Figure 1. The constructed domain in this study consisted of 100, 50 elements in x and y direction respectively.

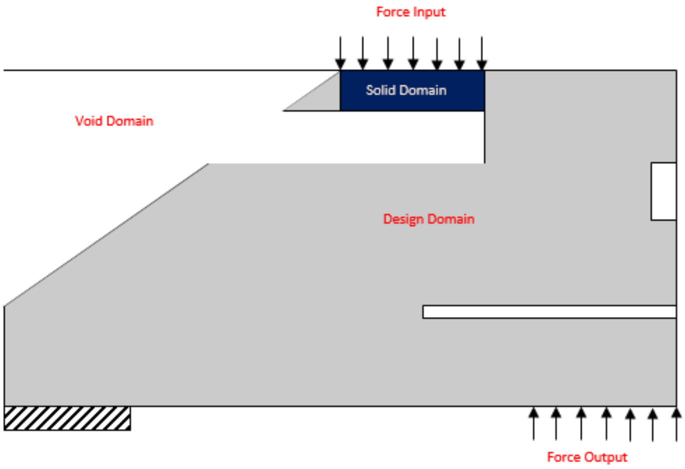

Figure 1: Initial Design, Domain

FEA analysis is performed in the initial design, domain, governed by the loading and constraint conditions assigned by the engineer. The first model of the required mechanism is developed in the design domain by discretizing the domain into a maximum number of 100 elements in x, 50 elements in my direction (nelx, Nely), as mentioned in the topology optimization program executed using MATLAB [17]. Dark Shaded regions in the design domain are considered to be solid regions and hence assigned a higher relative element density value. Solutions of the design domain for the specified boundary conditions are solved by the governing equations of FEA. Solid Isotropic Material with Penalization (SIMP) approach is used [18] to optimize the design domain. SIMP method assumes young's modulus 'E0' as '1' and uses penalization 'p' to make transitional densities 'ρ'.

$$E_i = \rho_i^p E_0 \tag{1}$$

In this research work, the percentage of material volume reduction is considered as 50% of the material from the initial volume. A mesh independent filter is used with filtering radius of 1.2 to remove the numerical instabilities such as check board patterns, mesh dependencies in the final solution.

The objective function of the topology optimization problem is formulated as given in equation (2).

$$
\begin{aligned}
-MPE = U_1^T K U_2 = \sum_{e=1}^{N} (x_e)^p \, (u_{e2}^T k_0 u_{e1}) \\
\text{Subject to} \quad : \quad V(x) / V_0 = f \\
: \quad [K]\{U\} = \{F\} \\
: \quad 0 < x_{min} \le x \le 1
\end{aligned}
\tag{2}
$$

Where:

{U}-Global displacement vectors, {F}-Global force vectors, [K]-Global stiffness matrix, u_{e1}-Element displacement vector due to input port, k_e - Stiffness matrix of the element, u_{e2}- Element displacement vector due to $f1_{out}$ output port, u_{e3} - Element displacement vector due to $f2_{out}$ output port, x - Vector of design variables, x_{min} - minimum relative densities (In the MATLAB program, non-zero value has been assumed to avoid singularity), N - Number of elements, P- Penalization power (typically p = 3), V(x)- Material volume, V_0-Design domain volume, f(volfrac)-Volume fraction.

In our research paper, only the heel strike loading condition was considered while generating the optimized model of required compliant prosthetic leg.

CAD Model Generation

Based on the configuration of the optimized model from MATLAB, the sketch is converted to a CAD model in SOLIDWORKS. The current Prosthetic foot designs do not consider the actual weight distributions of the human body along either foot. Qian et al has developed a computational framework for investigating the dynamic behavior and the internal loading conditions of the human foot during the gait cycle. We have hence incorporated the following feature by extruding the optimized CAD model along the typical pressure line of a human foot in order to study the variations in the analysis of deformation. This maximum pressure distribution is prevalent in majority average sample of the human population, weighing around 60 Kgs.

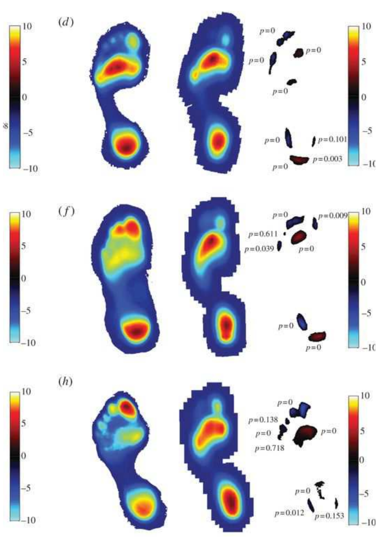

Figure 2: Pressure Distribution along Human Foot
http://d10k7sivr61qqr.cloudfront.net/content/royinterface/10/83/20130009/F3.large.jpg?width=800&height=600&carousel=1

Finite Element Analysis of the Developed Design

The Finite Element Analysis of the developed model was carried out in ANSYS Workbench. The parameters of deformation and equivalent Von-Mises stress were computed from the FEA model. The method of structural analysis was adopted from the simulation procedure followed by Anne Schmitz [19]. Foot and platens were created in SOLIDWORKS and exported to ANSYS work bench for analysis. To ensure feasible results, default structural steel material was used for simulating contours. However, change in contour domains will be observed on considering plastic materials such as ABS-grade plastics. (Material Properties of Structural Steel: Density: 7850 kg/m^3, Tensile Yield Strength: 2.5E+08 Pa, Tensile Ultimate Strength: 4.6E+08 Pa)

RESULTS AND DISCUSSIONS

The results for different stages in the methodology of the project have been tabulated and discussed below in a chronological sequence.

Topology Optimisation

The following optimized design of a compliant based prosthetic foot was generated after 200 iterations.

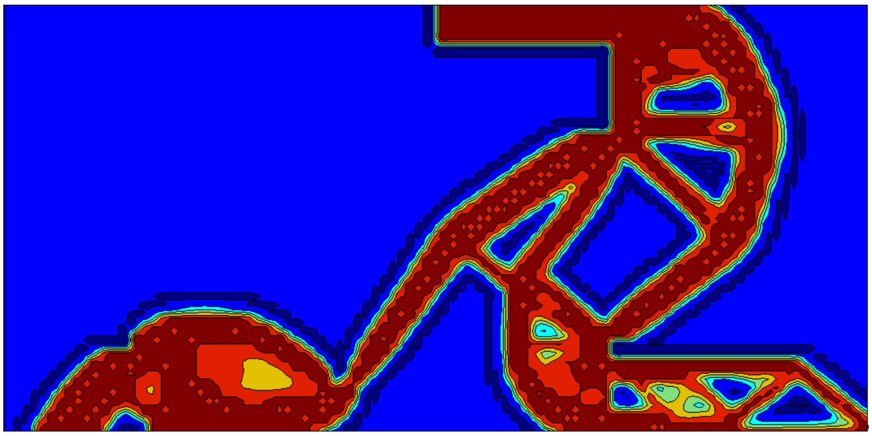

Figure 3: Topologically Optimised Design

The design was further exported to SOLIDWORKS and modelled to obtain the CAD model.

Cad Model

CAD Model was generated and extruded to a width of 100 mm, representing an average width of the human foot. The foot was further extruded along the pressure curve as described in figure.2 in order to increase the flexibility factor and yield more comfort to amputees in a variety of applications as discussed in previous literature surveys. The SACH foot involves no movement in the sub tarsal joint region. Hence no inversion is possible for amputees wearing this type of prosthetics. The SACH foot is only suitable for walking on even surfaces and completely unsuitable for use in rough terrains [20]. The JAIPUR FOOT on the other hands yields better comfort for amputees attempting to squat and walk on the similar surfaces.

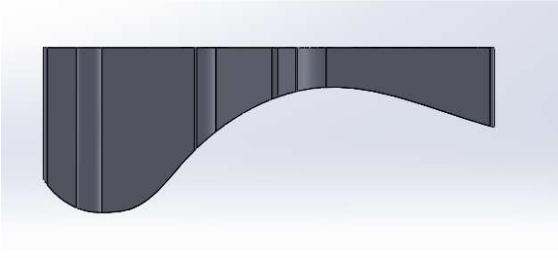

Figure 4: CAD Model Extruded along the Pressure Curve at the Base of the Developed Prosthetic Foot

The following represents the isometric view of the generated Prosthetic foot design. The model was further imported into ANSYS Workbench to perform structural analysis.

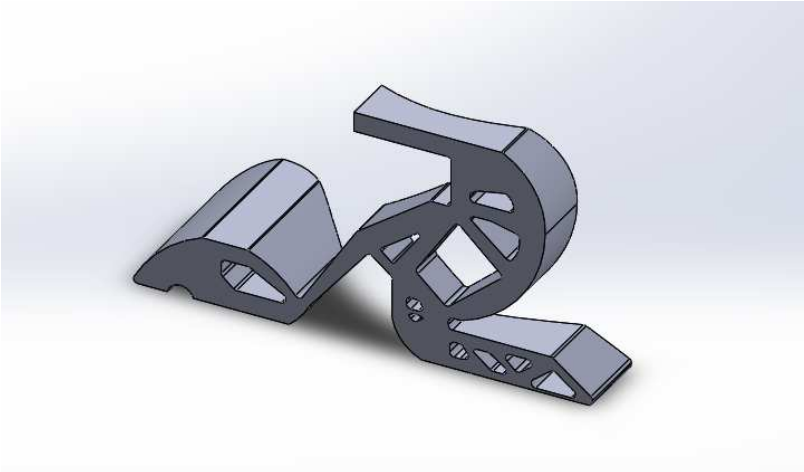

Figure 5: CAD Model Generated in SOLIDWORKS

Finite Element Analysis of the Designed Prosthetic Foot Model

Static structural analysis was first performed in ANSYS after suitable meshing. The three phases in a gait cycle were considered discreetly for static analysis. Modal analysis was further performed in the similar discrete method.

Static Structural Analysis

The platens were mated to the foot and meshed in ANSYS Workbench and initial loading conditions were provided to the foot. Furthermore, in the heel strike phase, the angle between the platen and foot was considered to be 15 degrees. A vertical load of 1600 N was applied to the foot considering a highest weight bearing scenario. Midstance phase was simulated by constraining the toe and heel completely to the platen. A vertical load of similar magnitude was provided to the foot, acting in the downward direction. Toe lift off phase was simulated by constraining the toe to one of the platen at an angle of 20 degrees. Furthermore, the similar vertical load was provided to the foot.

Total deformation and equivalent Von- Mises stress results for each of the three phases have been further described in chronology below. Lastly, the total life cycle of the product has been computed. The product can be certified as safe, the total life cycles before any unwarranted creep is 1e6.

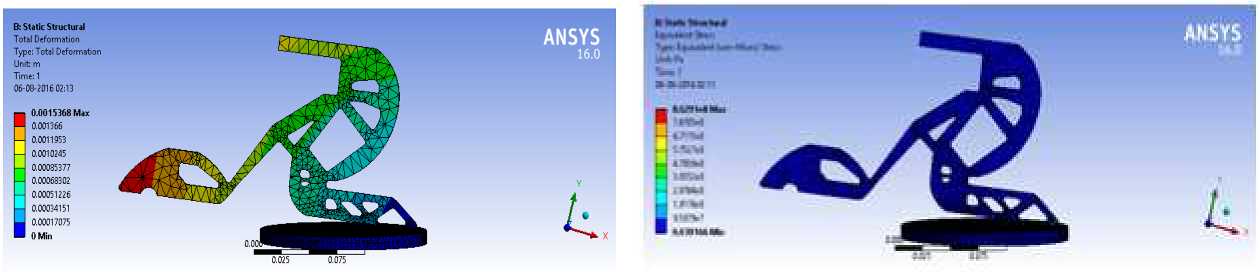

Figure 6: Total Deformation and Equivalent Stress Results in Heel Strike Phase

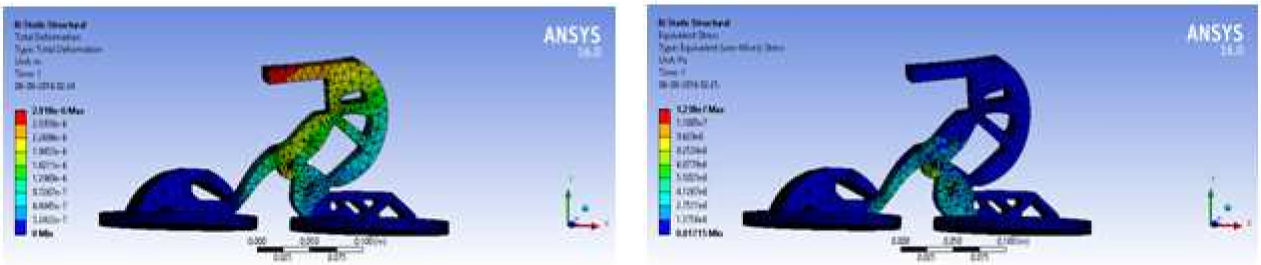

Figure 7: Total Deformation and Equivalent Stress Results in Mid Stance Phase

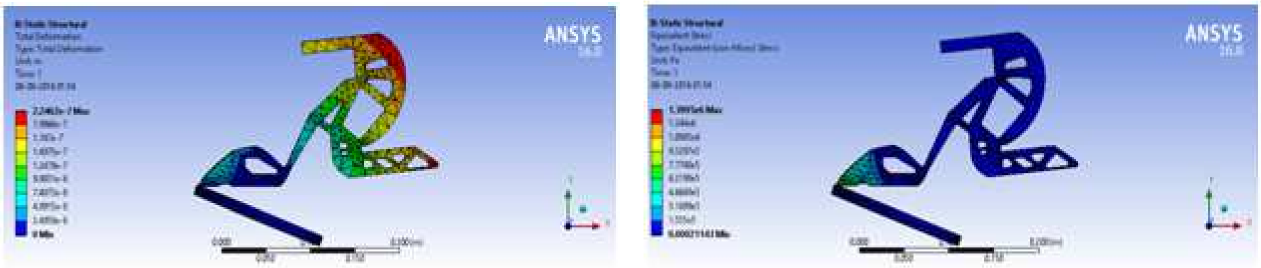

Figure 8: Total Deformation and Equivalent Stress Results in Toe Lift off Phase

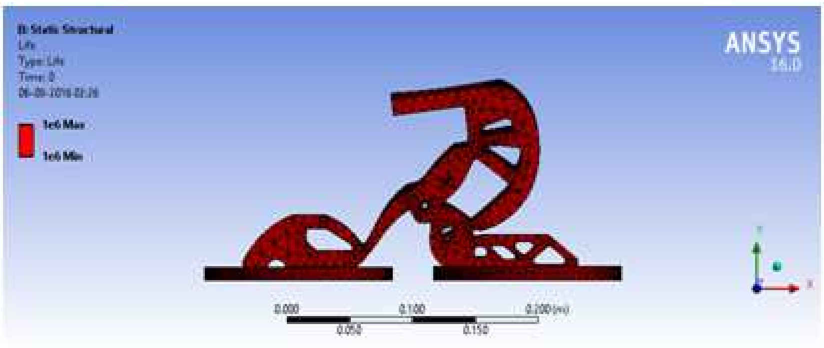

Figure 9: Life Cycle of the Developed Prosthetic Foot

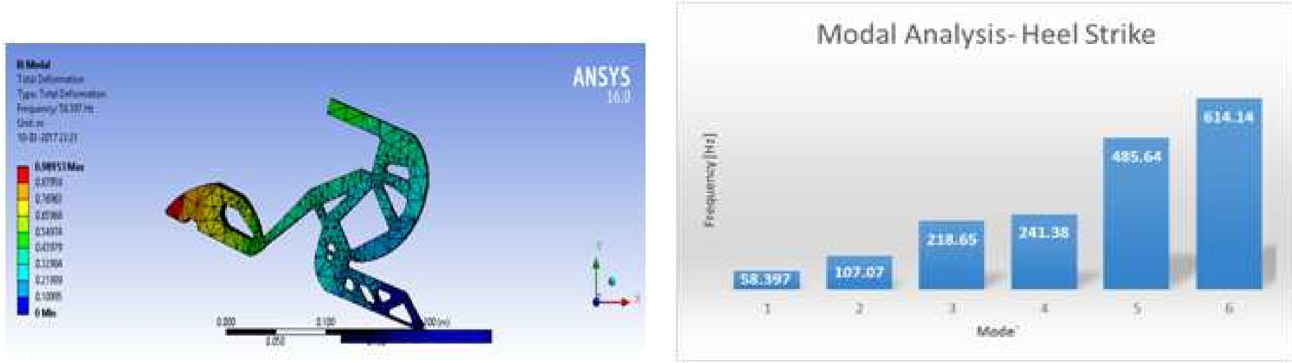

Figure 10: Total Deformation Modal Analysis Result in Heel Strike Phase

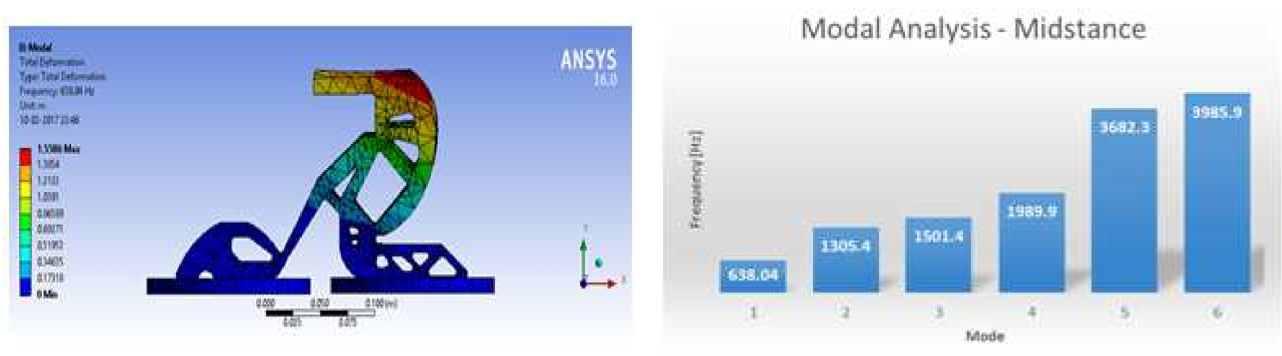

Figure 11: Total Deformation Modal Analysis Result in Mid Stance Phase

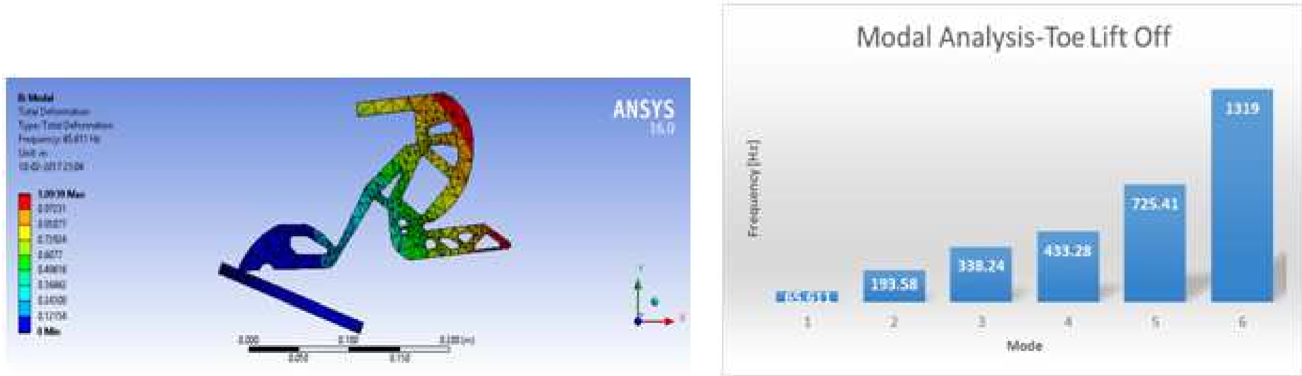

Figure 12: Total Deformation Modal Analysis Result in Toe Lift Off Phase

Modal Analysis

Modal analysis has been performed in order to obtain the mode shape results. The range of frequency for 6 mode steps have been computed for each of the three phases and tabulated below.

The deformation results advocate the negligible deformation and stress origination at the heel region at the instant of strike. The deformation tends to traverse to the toe of the developed foot through the elastic keel.

The FEA results in the midstance phase further points to a cumulative accretion of plastic deformation energy coupled by significant build of stress in the keel region. The propagation of deformation in the toe lift off phase further provides considerable evidence that our model is capable of delivering superior ESAR to the amputees during a periodic gait. The flexibility exhibited by our developed prosthetic foot is hence capable of supporting the amputees in challenging

tasks such as treading on gravel roads. The total life cycle is also determined to be safe for a maximum load of 120 KG. Hence the developed foot is certified safe for a maximum life cycle of 1e6 life cycle.

The modal analysis results helped us in establishing the natural frequency of the foot during different instances of the gait cycle. Since the natural frequency of the foot is relatively low in all the instances, the amputees will seldom encounter large dynamic oscillations. The periodic loading cycles will not coincide with the modal frequency hence circumventing oscillations at the resonance frequency.

Table 1: Tabulation of Results

Result Parameter	Heel Strike	Midstance	Toe Lift off
Total Deformation (m)	0.0015368	2.918e-6	2.2462e-7
Equivalent stress (Von Mises) (Pa)	8.6291e8	1.238e7	1.3995e6
Modal Frequency (Hz)	58.397	638.04	65.611

CONCLUSIONS

This research study presents a framework using topology optimization technique to develop new prosthetic foot with endorsing enhanced ESAR and flexibility. The framework was used to generate a prosthetic foot that minimized material storage but at the same time attempted to domain the acceptable stiffness characteristics and product life cycle. The final model was validated using FEA to verify the equivalent static stresses and modal responses. This developed framework can further be explored and implemented to fabricate prosthetics using rapid prototyping methods thus refining an amputee's gait cycle. The developed Prosthetic foot is robust and cost effective, inherently suitable for the Indian market.

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